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Impact of Wall Constraint on Velocity Distribution in Proximal Flow Convergence Zone

Implications for Color Doppler Quantification of Mitral Regurgitation

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Objectives. **This study sought to evaluate the effect of proximal flow constraint induced by the left ventricular wall on the accuracy of calculated flow rates and to assess a possible correction factor to adjust the proximal convergence angle. We further defined under which hydrodynamic and geometric conditions it is necessary to apply the corrected convergence angle.**

Background. **The proximal flow convergence method has been proposed as a new approach to quantify valvular regurgitation. However, significant overestimation of the calculated regurgitant flow rate has been reported, particularly in patients with mitral valve prolapse and severe mitral regurgitation.**

Methods. **We used an in vitro flow model and induced various degrees of proximal flow constraint. The accuracy of the proposed convergence angle formula, $\alpha = \pi + 2\tan^{-1} d/r$ (d = wall distance; r = isovelocity radius) was tested in vitro and in a three-dimensional numerical simulation.**

Results. **With a constraining wall near the orifice, overestimation of regurgitant flow rates was noted and was most significant with the constraining wall positioned closest to the orifice (calculated flow rate [Q_c]/true flow rate [Q_o] = 1.85 ± 0.55 [mean ± SD]). These findings were similar to the results of the numerical simulation. Applying the correction factor nearly completely eliminated the overestimation of the calculated flow rates (cQ_c), with cQ_c/Q_o = 1.13 ± 0.25.**

Conclusions. **In the presence of a constraining wall, significant overestimation of calculated flow rates is observed when hemispheric symmetry of the flow field is assumed. In this situation, it is necessary to apply the corrected convergence angle formula to improve the accuracy of the proximal flow convergence method.**

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Two-dimensional color Doppler flow mapping provides a convenient and noninvasive way to visualize normal and abnormal blood flow in the cardiac chambers (1). In the presence of valvular regurgitation, a mosaic color pattern is seen in the receiving chamber. Initially, it was hoped that this method would provide an easy and accurate way to quantify valvular regurgitation based on simple measurements of length, width or area of the regurgitant jet (2–4). However, it was soon recognized that the size of the regurgitant jet displayed in the color map depends on multiple physical, geometric and instrument factors other than regurgitant flow rate itself (5–7).

Recently, a new method for quantification of valvular regurgitation was introduced based on the conservation of mass in the proximal converging flow field (8). This method has been shown to be accurate using numerical and in vitro flow modeling; however, controversy remains about the accuracy of this approach in the clinical situation. We and several investigators have found promising results in patients with moderate and moderately severe mitral regurgitation (9,10). However, significant overestimation of the mitral regurgitant flow rate has been found, particularly in patients with mitral leaflet flail and severe regurgitation (11). Although a few possible mechanisms have been postulated, none of these has been studied in a systematic way to explain these clinical observations satisfactorily (12–14).

We hypothesize that the converging flow field is distorted by proximal constraint imposed by valvular structures and the left ventricular wall in patients with distorted mitral valve anatomy and severe regurgitation. The assumption of hemispheric symmetry of the proximal flow convergence zone in this situation will lead to significant overestimation of regurgitant flow rates. In this study we evaluate the impact of proximal flow constraint on the velocity distribution in the converging flow field and the accuracy of calculated flow rates assuming hemispheric symmetry in an in vitro flow model and a three-dimensional computational fluid dynamics simulation. We propose and validate a possible correction factor to improve the accuracy of flow rate calculations in the presence of varying degrees of proximal flow constraint.

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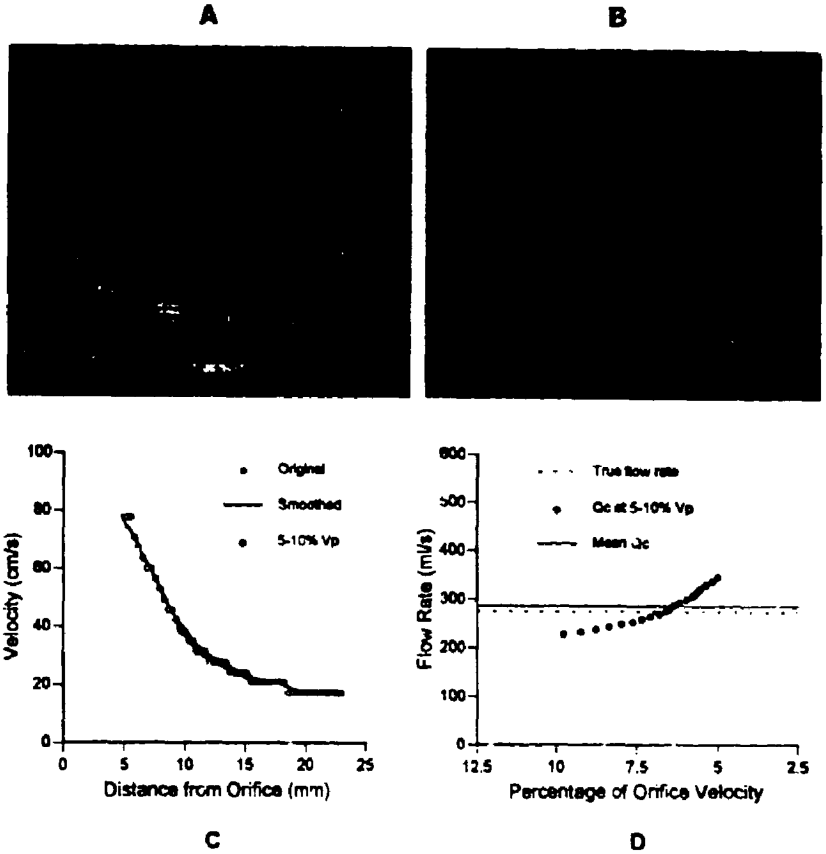

Figure 1. Analysis of color Doppler flow images. **A,** Two-dimensional color Doppler flow image (as displayed on the echocardiogram) of the proximal flow convergence zone obtained in the in vitro experiment. **B,** Color-coded display of the averaged velocity field of 31 consecutive color Doppler flow images (displayed on the Macintosh). The centerline (**white line**) is determined from the orifice location (identified on the black-and-white image) to the periphery of the flow field. **C,** Raw centerline velocity profile (**open symbols**) with velocities on the Y-axis and distance from the orifice on the X-axis. The **solid line** illustrates the smoothed velocity profile, and the **solid symbols** identify all centerline velocities between 5% and 10% of the orifice velocity, used in the subsequent flow calculations. **D,** Illustration of the calculated flow rates (Y-axis) obtained at all the selected centerline velocities (X-axis). The true flow rate (**dashed line**) and mean calculated flow rate (**solid line**) are superimposed.

Methods

In vitro study. For this study we used an in vitro flow system previously described (15). Four different-sized circular orifices with areas of 0.1, 0.3, 0.5 and 1.0 cm^2 were tested. Steady-state orifice flow rates ranged from 32 to 280 cm^3/s with orifice velocities between 310 and 510 cm/s to simulate clinically relevant hydrodynamic conditions. True orifice flow rate (Q_o) was measured by timed collections. At baseline, flow rate was evaluated without proximal constraint, allowing flow convergence over 2π radians (the geometry around the orifice was planar). Subsequently and for the same hydrodynamic conditions, flow constraint by the ventricular wall was simulated by attaching a curved plastic wall with internal diameter of 50 mm and an arc of 90° at 1, 3, 5 and 10 mm distant from the orifice edge.

Echocardiographic data. The proximal flow convergence area is imaged using a Sonos 1500 (Hewlett-Packard) equipped with a 2.5-MHz transducer positioned along the centerline of the flow. The color Doppler sector is minimized to optimize frame rate. The lowest available wall filter setting is used, and spatial filters are turned off. Color gain is adjusted to avoid color artifacts. Thirty consecutive two-dimensional color Doppler flow images are captured in a cine loop and stored to optical disk in Tagged Image File Format-related format for off-line analysis (Fig. 1). Continuous-wave Doppler is used to measure peak orifice velocity (v_p).

Data analysis. Customized software is written in LabView II (National Instruments) using an additional library of image-processing tools, Ultimage Concept VI (Graftek). All digitally stored images are baseline-shifted and averaged for further analysis (Fig. 1B). Baseline shifting provides us with a wider range of unaliased forward velocities from 0 to twice the Nyquist velocity. The color assignments used for the computer display of the averaged images is randomly chosen to highlight each velocity contour and does not necessarily match the color map of the echocardiograph (Fig. 1A,B). The coordinates of the orifice location are identified on the black-and-white image and are used to draw the centerline on the averaged velocity field from the orifice to the periphery of the flow convergence (Fig. 1B). Centerline velocity profiles are reconstructed and low-pass filtered (first-order Butterworth filter with 0.3 pixels/mm cutoff frequency applied forward and reverse) to remove high-frequency noise (Fig. 1C). Regurgitant flow rate Q_c is calculated simultaneously at consecutive isovelocity shells between 5% and 10% of the

JACC Vol. 27, No. 3
March 1, 1996:706-13

orifice velocity assuming hemispheric symmetry of the converging flow field as (Fig. 1D):

$$Q_c = 2\pi r^2 v,$$

where r is the distance from the orifice to the isovelocity shell and v the corresponding velocity (Fig. 1D). All flow calculations along the centerline are averaged and reported as a mean calculated flow rate ± SD. They are compared with true flow using linear regression analysis. The ratio between calculated and true flow (Q_c/Q_o) and the difference ($\Delta Q = Q_c - Q_o$) is obtained and expressed as a mean ratio ± SD. We have previously shown by numerical simulations and in vitro flow modeling that at isotachs between 5% and 10% of the orifice velocity, the flow field converging toward a finite circular orifice with planar surroundings closely approximates the hemispheric velocity distribution of inviscid flow converging toward a point orifice (16). This range of proximal velocities selected for analysis also corresponds closely to the optimal velocity thresholds recently proposed by other investigators (17). It has been shown that orifice flow rate is underestimated when calculated at isovelocity shells too close to the orifice because progressive flattening of the isotachs occurs as flow approaches closer to a finite orifice (16,17). This underestimation increases approximately linearly with the increase of isotach/orifice velocity ratio (v_a/v_p). Therefore, velocities >10% of the orifice velocity were excluded from subsequent analysis (even if they were displayed unambiguously). Velocities <5% of the orifice velocity were discarded because the presence of confounding flow fields, frame-to-frame variability, limited low-velocity resolution and the instrument wall filter can significantly affect the accuracy of the lower velocities displayed in the converging flow field.

Because, in the clinical situation, analysis of the proximal flow field is performed on individual two-dimensional color Doppler flow images, we selected for three hydrodynamic stages five individual frames showing a homogeneous proximal flow convergence zone with minimal color noise. Flow rate was calculated using the previously described steps on the individual frames to evaluate the effect of frame-to-frame variability on the accuracy of the flow calculations based on the selected range of velocities. The individual values were compared with true flow and flow calculated from the averaged images. Q_c/Q_o and ΔQ were calculated and expressed as the mean value ± SD.

Correcting the proximal convergence angle. When flow rate calculations are performed on a single isovelocity shell, the convergence angle (α) can be simply measured on the two-dimensional color Doppler image using a protractor. We have recently validated this method in the intraoperative setting using transesophageal echo (18). However, if flow calculations are performed simultaneously at multiple isovelocity shells on the same image (as is done with this automated computer algorithm) the convergence angle (α) should be measured for each of these isovelocity shells individually. As this would be nearly impossible in a clinical setting, we propose to approximate the convergence angle (α) for each isovelocity shell as a

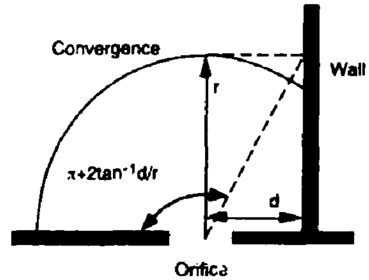

Figure 2. Calculation of the corrected convergence angle (α) in the presence of a nearby constraining wall.

function of wall distance d and proximal velocity contour at radius r from the orifice (Fig. 2):

$$\alpha = \pi + 2\tan^{-1}\frac{d}{r}.$$

For example, a nearby wall (at distance d) will exert more flow constraint at isovelocity shells further from the orifice (larger r) than at isotachs close into the orifice (smaller r). When the distance d between the wall and the regurgitant orifice is measured and entered in the computer algorithm, the convergence angle α for each isovelocity contour is calculated automatically, and the corrected flow rate (cQ_c) is calculated using the adjusted convergence angle as follows:

$$cQ_c = \left(\pi + 2\tan^{-1}\frac{d}{r}\right)r^2 v.$$

The corrected calculated flow rates (cQ_c) are reported as a mean calculated flow rate ± SD. The ratio between corrected calculated flow and true flow (cQ_c/Q_o) and the difference ($\Delta cQ = cQ_c - Q_o$) are obtained and expressed as a mean ratio ± SD. These data are compared with the uncorrected estimates by paired *t* test and analysis of covariance.

From this formula it can be readily derived that when d ≤ r, $\tan^{-1}$ d/r ≤ $\pi/4$, and thus the flow convergence angle will be limited to $\alpha \le 3\pi/2$. If in this case hemispheric flow convergence is assumed ($\alpha = 2\pi$), the calculated flow rate will overestimate true flow by at least 25%. Therefore, to adjust for this overestimation, the corrected convergence angle formula is applied in the situation when d ≤ r. However, when d > r, $\tan^{-1}$ d/r will be closer to $\pi/2$, and thus the convergence angle α will approximate 2π ($\alpha \cong 2\pi$) indicating flow convergence over nearly a full hemisphere (no significant constraint). In this case the calculated flow rates assuming hemispheric flow convergence will closely approximate true flow. Therefore, the convergence angle correction is not applied for isovelocity contours where d > r.

Numerical simulation. To evaluate further the impact of proximal flow constraint and the accuracy of the proposed correction factor, we perform a series of three-dimensional numerical simulations using a commercially available finite difference program (Fluent V4.25) running on a Silicon Graphics Iris Indigo workstation. Calculations are based on a

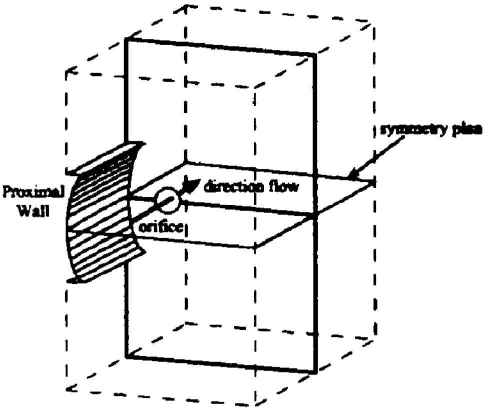

Figure 3. Three-dimensional geometry used for the numerical simulations.

99,216-node model (6.75-cm width, 5.35-cm height, 10-cm depth) with a plane of symmetry perpendicular to the plane of the orifice (Fig. 3). An orthogonal computational grid is used, and the size of the individual cells is 1 mm^3 for a 2.5-cm radius surrounding the orifice, gradually increasing to a cell size of 6.8 mm^3 at the outer boundaries of the model. The proximal geometry includes a 90° sector of a curved wall with a 5-cm diameter located at 1 and 10 mm from the orifice edge. For these two geometries, steady flow through a 0.48 cm^2 round orifice is simulated for three hemodynamic conditions. Noncompressible fluid with physiologic density and viscosity are used (1.04 g/cm^3 and 3 cP, respectively). Mass flow rate is conserved at inlet and outlet. Convergence criteria are set at a relative residual error of less than 3×10^{-3} (~1,000 iterations). Similar to the in vitro study, flow rate is calculated using the centerline velocities between 5% and 10% of the orifice velocity assuming hemispheric symmetry of the converging flow field. To validate the proposed correction factor, a corrected flow rate is calculated when the distance (d) to the constraining wall is smaller than the radius (r) of the isovelocity shells and compared with orifice flow rate. When $d > r$, no correction is applied.

Results

Numerical simulation. Figure 3 shows the geometry used for the simulation illustrating the orifice and the adjacent constraining wall. Figure 4 illustrates a color-coded velocity magnitude and u-velocity distribution (component parallel with orifice flow) of two orthogonal cross sections through the proximal flow field in the presence of a constraining wall at 1 mm from the orifice. It shows the distortion of the proximal isovelocity shells imposed by the nearby wall. With the constraining wall at 1 mm, assuming hemispheric flow convergence causes significant overestimation of the orifice flow rate with $Q_c/Q_o = 1.65$. Because the distance (d) to the wall is smaller then the radii (r) of the isovelocity shells analyzed ($d \leq r$), the correction factor to adjust the convergence angle is applied, and a corrected flow rate (cQ_c) is calculated. Applying this correction factor nearly completely corrects the overestimation of orifice flow rates with $cQ_c/Q_o = 1.03$. With the proximal wall further away from the orifice ($d > r$), assuming hemispheric flow convergence yields accurate flow rate calculations with $Q_c/Q_o = 1.02$, and no correction is applied.

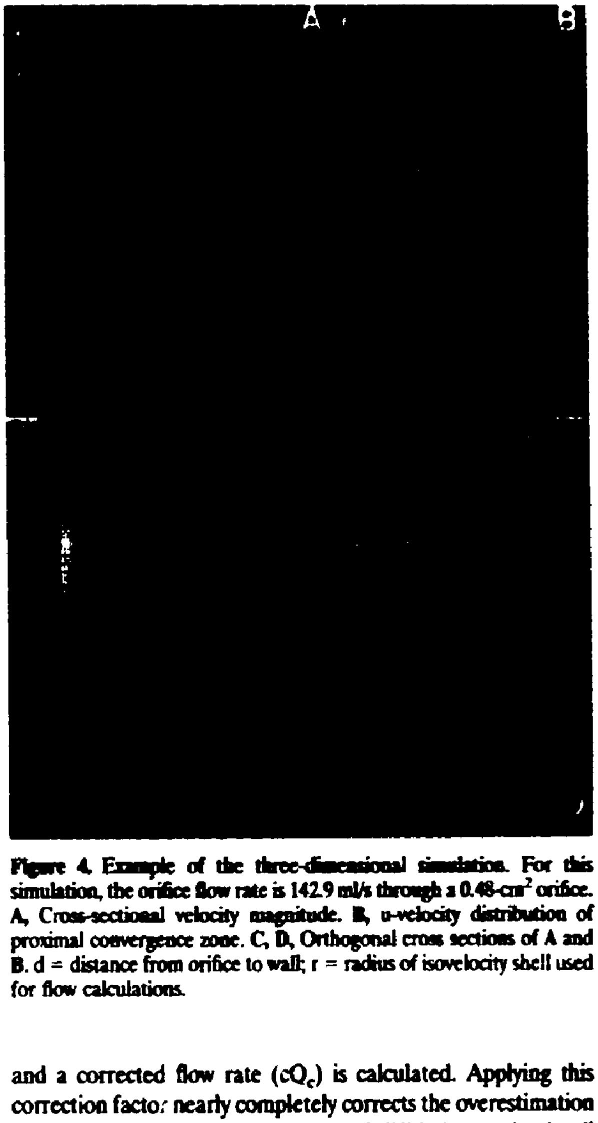

Figure 4. Example of the three-dimensional simulation. For this simulation, the orifice flow rate is 142.9 ml/s through a 0.48-cm^2 orifice. **A,** Cross-sectional velocity magnitude. **B,** u-velocity distribution of proximal convergence zone. **C, D,** Orthogonal cross sections of A and B. d = distance from orifice to wall; r = radius of isovelocity shell used for flow calculations.

In vitro study. Figure 5A illustrates a typical example of a velocity-distance profile along the centerline of the proximal convergence zone. The dotted segment of the proximal velocity-distance profile indicates the velocities between 5% and 10% of the orifice velocity. The open symbols represent a situation without proximal flow constraint. The solid symbols represent the velocity-distance profile for the same hydrodynamic condition with a constraining wall at 3 mm from the orifice. Notice that for the same hydrodynamic condition, the entire velocity-distance profile is shifted upward and to the right in the presence of proximal flow constraint. Figure 5B shows the corresponding calculated flow rates at all selected centerline velocities. Without proximal flow constraint, the calculated flow rates at all individual isovelocity shells (open

JACC Vol. 27, No. 3
March 1, 1996:706-13

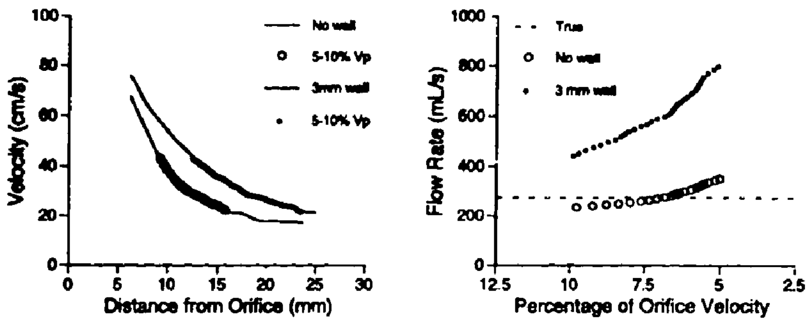

Figure 5. Impact of proximal flow constraint on centerline velocity-distance profile and calculated flow rates assuming hemispheric symmetry. **Left,** Centerline velocity profiles and, superimposed, the velocities between 5% and 10% of the orifice velocity without proximal flow constraint (**open symbols**) and with the wall positioned at 3 mm from the orifice (**closed symbols**) for a constant flow rate. **Right,** Corresponding calculated flow rates; the true orifice flow is superimposed as a **dashed line.**

symbols) are close to the true flow represented by the dotted line. In the presence of proximal flow constraint, however, the calculated flow rates (solid symbols) significantly overestimate true flow. Also notice the large difference between calculated flow rates at lower velocities (5% of the v_p) and higher velocities (10% of v_p), because the lower velocity contours, being further from the orifice, are affected more by proximal flow constraint.

In the absence of proximal flow constraint, the calculated flow rate assuming hemispheric symmetry (Q_c) closely approximates true flow rate (Q_o) throughout a wide range of hydrodynamic conditions, with $Q_c/Q_o = 1.03 \pm 0.08$ (range 0.94 to 1.17), $\Delta Q = 10 \pm 19$ ml/s and $r = 0.99$. The presence of proximal flow constraint imposed by the ventricular wall at 10 mm from the orifice causes only minimal overestimation of the calculated flow rate ($Q_c/Q_o = 1.11 \pm 0.17$ and $\Delta Q = 25 \pm 39$ ml/s). This overestimation is present only for the higher flow rates through the larger orifices (0.5 and 1 cm^2); flow through the smaller orifices (0.1 and 0.3 cm^2) is accurately calculated. As the constraining wall is positioned closer to the orifice, the overestimation gradually increases. With the wall positioned at 1 mm from the orifice edge, the calculated flow rate overestimates true flow significantly with $Q_c/Q_o = 1.85 \pm 0.55$ range 1.16 to 2.71) and $\Delta Q = 154 \pm 173$ ml/s. In this case, overestimation is present for all orifice sizes and all flow rates.

Evaluation of frame-to-frame variability. To evaluate the effect of frame-to-frame variability on the accuracy of the centerline velocity-distance analysis, five single images are selected for three hydrodynamic conditions without wall constraint. The images are selected based on the appearance of a hemispheric flow convergence zone without color noise interference. The calculated flow rate from the individual color Doppler images closely approximates true flow with $Q_c/Q_o = 1.03 \pm 0.07$ (range 0.91 to 1.18) and $\Delta Q = 6 \pm 10$ ml/s.

Evaluation of correction factor. In the presence of proximal flow constraint, the distance d between the regurgitant orifice and the wall is measured and compared with the radius r of all the isovelocity shells selected for analysis for each hydrodynamic condition. When $d \le r$, the correction of the convergence angle is applied to the flow rate calculations (cQ_c). For the largest orifice (1 cm^2), the convergence angle correction is necessary for all degrees of wall constraint. For the 0.5 cm^2 and 0.3 cm^2 orifices, angle correction was necessary for all hydrodynamic conditions except for a wall distance of 10 mm. For the smallest orifice (0.1 cm^2), convergence angle correction was necessary only with the wall positioned very close to the orifice (1 mm and 3 mm). Overall, when $d \le r$, $Q_c/Q_o = 1.64 \pm 0.50$ and $\Delta Q = 125 \pm 150$ ml. Applying the corrected convergence angle in these conditions reduced the overestimation of calculated flow rates significantly ($p < 0.001$), with $cQ_c/Q_o = 1.13 \pm 0.25$ (range 0.90 to 1.63) and $\Delta cQ = 37 \pm 67$ ml/s (Fig. 6). Only for the largest orifice was the overestimation not completely corrected. When $d > r$, $Q_c/Q_o = 1.05 \pm 0.11$, and no correction was applied. All results are summarized in Table 1.

Discussion

Over recent years Doppler echocardiography has become the method of choice for serial evaluation of both valvular structures and hemodynamics. Currently, color Doppler assessment of regurgitant lesions is predominantly semiquantitative, based on simple measurements such as height, length or area of the regurgitant jet displayed by color flow mapping (2-4). Recently, the proximal flow convergence method has

Figure 6. Ratio of calculated to true flow rate (Q_c/Q_o) in the presence of significant flow constraint ($d \le r$) assuming hemispheric flow convergence (**left**) and applying the corrected convergence angle formula (**right**).

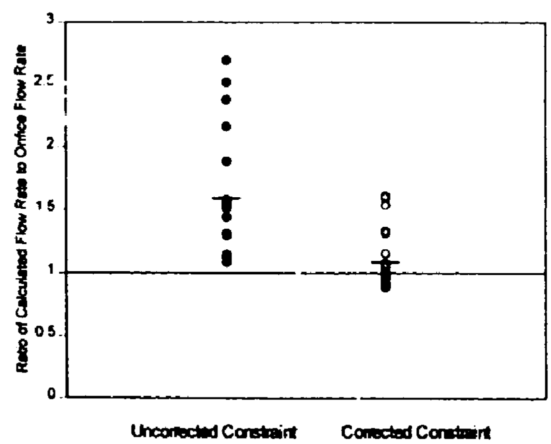

Table 1. Evaluation of Correction Factors for Flow

Wall Distance (mm)	Orifice Size (cm^2)	V_o (cm/s)	Q_o (ml/s)	Q_c/Q_o	cQ_c/Q_o	Radius Range (mm)	Convergence Angle α (degrees)
1	0.1	400	31	1.16	0.97	4.6-6.9	224-244
3			31	1.13	0.94	4.1-5.8	259-266
5			31	1.03	NA	3.4-5.8	360
10			31	0.97	NA	3.2-5.8	360
No wall			32	0.94	—	3.2-5.8	360
1	0.3	400	81	1.58	1.03	6.5-13.0	220-244
3			81	1.31	0.90	5.8-10.3	241-269
5			80	1.10	1.01	6.2-9.6	262-272
10			81	0.91	NA	6.2-9.3	360
No wall			80	1.04	—	5.8-8.2	360
1	0.5	310	104	1.55	0.98	6.9-14.4	218-251
3			106	1.33	0.93	6.5-13.4	235-266
5			106	1.15	0.92	6.5-13.0	249-273
10			104	1.05	NA	6.5-12.1	360
No wall			105	0.98	—	6.2-11.0	360
1	0.5	400	117	1.91	1.17	9.3-16.1	214-232
3			148	1.46	0.97	8.2-14.7	229-262
5			150	1.31	0.94	7.9-13.7	247-272
10			146	1.10	NA	7.2-12.3	360
No wall			150	1.01	—	7.2-11.7	360
1	0.5	510	174	2.18	1.33	9.6-16.8	212-236
3			182	1.60	1.06	7.9-15.1	229-262
5			180	1.53	1.09	7.9-14.4	243-268
10			181	1.25	NA	7.5-13.0	360
No wall			184	1.06	—	7.2-11.7	360
1	1.0	440	280	2.71	1.63	13.4-25.7	209-233
3			275	2.53	1.62	11.7-24.7	218-253
5			270	2.39	1.56	11.3-24.3	227-266
10			270	1.59	1.35	9.6-19.2	258-269
No wall			275	1.17	—	8.9-17.5	360

The range of radii corresponds to the isovelocity shells between 5% to 10% of the orifice velocity used for the analysis. Convergence angle α represents the solid angle in degrees. Without proximal constraint, $\alpha = 2\pi = 360°$. The range of α represents the range of convergence angles calculated for different radii r corresponding to the isovelocity shells where $d < r$. cQ_c = corrected calculated flow rate; NA = not applied; Q_c = calculated flow rate (no correction); Q_o = orifice flow rate; V_o = orifice velocity.

shown promise in the noninvasive quantitation of valvular regurgitation (8). This method allows us to calculate regurgitant flow rate (9) and effective regurgitant orifice area (19) based on the principle of conservation of mass in the proximal flow field. However, using the proximal flow convergence method in some patients with severe mitral regurgitation resulted in calculated regurgitant flow rates that are much higher than what would be expected based on the pathophysiology of valvular regurgitation (10,11).

The current implementation of the proximal flow convergence method inherently assumes hemispheric symmetry of the velocity distribution proximal to the regurgitant lesion. This assumption is true for inviscid flow approaching an infinitesimal orifice with planar surroundings (16). However, we have previously addressed some limitations where this assumption may no longer be valid, such as close to a finite orifice, where flattening of the isotachs occurs (16), or in the presence of nonplanar surroundings as seen in mitral stenosis (20).

In the present study we investigated in a systematic way the effect of solid boundaries, such as the left ventricular wall adjacent to the regurgitant orifice, on the velocity distribution in the accelerating flow field and subsequently on the accuracy of regurgitant flow rate calculations. We determined when an adjacent wall is causing proximal flow constraint and we propose a possible correction factor to improve the accuracy of flow rate calculations under these conditions.

Results of the current study. We and several other investigators have previously shown that assuming hemispheric symmetry of the proximal flow convergence zone is most accurate in the determination of flow rate calculations through circular orifices with planar surroundings (8,15). This is confirmed in this study, where we find accurate flow rate calculations by the proximal flow convergence method in the absence of flow constraint in the in vitro experiments. In the presence of wall constraint adjacent to the regurgitant orifice, the isovelocity shells cannot develop as full hemispheres; they are distorted and forced away from the orifice because of the principle of conservation of mass. This is confirmed by the results of the numerical simulations and by the in vitro findings, where an upward and rightward shift of the velocity-distance profiles is observed in the presence of proximal flow constraint (Fig. 5). This shift in the velocity-distance profile leads to significant overestimation of the calculated flow rates when hemispheric symmetry of the converging flow field is assumed. The distortion of the flow field is most prominent when the constraining wall is positioned closest to the orifice, which is also reflected in the accuracy of the calculated flow rates showing increasing overestimation when the wall constraint is closer to the orifice. The impact of the flow constraint imposed by the proximal wall is also more pronounced in the presence of higher flow rates and at the isovelocity shells further away from the orifice.

To improve the accuracy of the calculated flow rates obtained by the proximal flow convergence method in the presence of proximal flow constraint, we propose a correction factor to account for the reduction in convergence angle imposed by the constraining wall. With planar orifice surroundings, flow converges over 2π steradians; with axisymmetric constraint, the convergence angle is $2\pi(1 - \cos\theta)$, where θ is ½ center angle (15); in the presence of a unidimensional wedge constraint, the correction factor is $2\pi(\alpha/180)$, where α is the convergence angle (15,20). We believe that the constraint imposed by the ventricular wall is more unidimensional than axisymmetric, but it is not a simple wedge. If flow rate is calculated at a single isovelocity shell highlighted as the blue-to-yellow aliasing contour in the color Doppler velocity map, the convergence angle can be simply measured using a protractor. We have recently shown that this approach significantly improves the accuracy of the calculated flow rates, in particular in patients with posterior leaflet prolapse or flail (18). However, to automate the analysis and obtain more robust flow calculations, we use more of the centerline velocity information available in the Doppler maps.

Because it is not possible to measure convergence angles for each isovelocity contour individually, we propose to adjust the convergence angle for the different isovelocity shells as a function of distance (d) between the regurgitant orifice and the ventricular wall and radius (r) of the isovelocity contour using the following relationship: $\alpha = \pi + 2\tan^{-1}d/r$. This corrected convergence angle function is first tested in a three-dimensional numerical simulation. We find that in the presence of a nearby wall, this correction factor nearly completely eliminates the overestimation of calculated flow rates observed when hemispheric flow convergence was assumed. We also find in our numerical simulation that when the distance to the wall is larger than the radius of the isovelocity shell of interest, the amount of overestimation in calculated flow assuming hemispheric symmetry is very limited. Similar observations are made in the in vitro experiments; for all cases were $d > r$, calculated flow rates are not significantly different from the situation without wall constraint. Therefore, if $d > r$, adjustment of the convergence angle 2π in the subsequent flow rate calculations appears not to be necessary. If the wall distance is equal to or smaller than the isovelocity radius ($d \leq r$), the adjacent wall may cause distortion of the isovelocity shells, leading to significant overestimation of the calculated flow rates assuming hemispheric symmetry of the proximal flow field. Therefore, if $d \leq r$, application of the adjusted convergence angle $\alpha = \pi + 2\tan^{-1} d/r$ is necessary to correct the overestimation and improve the accuracy of the calculated flow rates significantly.

The findings of this study have important clinical implications. It provides a potential explanation for the overestimation observed by several investigators in some patients, therefore questioning the fundamental ability and accuracy of the proximal flow convergence method to quantify the severity of valvular regurgitation. In the absence of any obvious flow constraint, the proximal flow convergence method can provide an accurate quantitative estimate of the severity of a regurgitant lesion (21). However, when the proximal convergence zone is adjacent to the ventricular walls, assumption of a hemispheric isovelocity surface area will result in significant overestimation of calculated flow rate. This study also finds that a proximal wall does not significantly distort the proximal flow field if the distance (d) between the proximal wall and the regurgitant orifice is larger than the radius (r) of the isovelocity shell analyzed, and therefore, if $d > r$, the hemispheric assumption of the proximal flow field is still valid in the subsequent flow rate calculations. A proximal wall will induce flow constraint when it is closer to the regurgitant orifice than the radius of the isovelocity shell of interest ($d \leq r$). In this case, we propose a correction factor to adjust the proximal convergence angle α based on simple measurements of wall distance d and isovelocity radius r. Both of these measurements can be easily obtained from two-dimensional echocardiographic images. When the corrected convergence angle is applied in the subsequent flow rate calculations, the accuracy of the calculated flow rates using the proximal flow convergence method improves significantly.

This study is performed on averaged images (30 frames) to minimize random noise; however, in the clinical setting the proximal flow convergence zone is analyzed on single frames. Analysis of a subset of individual frames also showed excellent results, thereby lending credibility to this analysis technique and supporting its potential in the clinical situation.

Limitations and future directions. This study points out one of the potential pitfalls in the clinical application of the proximal flow convergence method. Proximal flow constraint may not be the only potential factor affecting the accuracy of the proximal flow convergence method (21). We and other investigators have previously addressed the effects of orifice size (16) and geometry (22,23), valvular geometry (13,20), leaflet motion (24), presence of concomitant outflow (14) and instrument factors such as aliasing velocity (16) or wall filter setting (12). Dynamic changes in regurgitant orifice area during the cardiac cycle may also result in over- or underestimation of regurgitant flow rate by the proximal convergence method (25). However, in a recent clinical study, we found that in patients with an eccentric convergence zone close to the posterior left ventricular wall, proximal flow constraint accounted for more than 75% of the overestimation in calculated regurgitant flow rate and regurgitant orifice area (18).

The correction factor proposed in this study adjusts the proximal flow convergence angle based on the relation between wall distance d and isotach radius r. Overall, this correction factor provides good accuracy of the calculated flow rates. Only for the largest orifice with the constraining wall closest to the orifice did the proposed correction factor not completely correct the overestimation. This may be related to the curvature in our constraining wall (instead of a flat wall) inducing more distortion of the converging flow field in the presence of very high flow rates than is being accounted for by the corrected convergence angle. Although further study is needed, this may not represent a significant limitation in the clinical setting because patients with a 1.0-cm^2 regurgitant orifice area and such high regurgitant flow rates have severe mitral regurgitation and may not pose a diagnostic or management dilemma. More importantly, in the range of moderate and moderately severe regurgitation, the proposed correction of the convergence angle provides highly accurate calculated flow rates, indicating the great potential of this correction factor in the clinical setting.

Although the proposed correction factor may appear complex and cumbersome, it is easily calculated from simple measurements obtained from the two-dimensional echocardiographic images. In the future, however, this correction factor could easily be incorporated into computer algorithms that could automatically calculate regurgitant flow rate, regurgitant orifice area and even correct for the presence of proximal flow constraint.

Conclusions. This study confirms the fundamental ability of the proximal flow convergence method to provide an accurate quantitative assessment of the severity of valvular regurgitation. However, in the presence of proximal flow constraint, distortion of the converging flow field can lead to significant overestimation of calculated regurgitant flow rates. Overestimation is more pronounced in the presence of higher flow rates or when the constraining wall is closer to the regurgitant orifice. These findings can explain the unexpectedly high regurgitant flow rates previously reported. In this study we define when an adjacent wall will induce proximal flow constraint and overestimation of the calculated flow rat s. We also propose a correction factor that allows accurate flow rate calculations throughout a wide range of hydrodynamic conditions with varying degrees of proximal flow constraint. This work and the proposed correction of the proximal convergence angle should allow the application of the proximal flow convergence method throughout a wider range of clinically relevant situations, in particular in patients with significant mitral regurgitation and a proximal convergence zone very close to the posterior ventricular wall, as commonly seen in patients with posterior leaflet prolapse or posterior leaflet flail.

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